Thin, flexible PDMS microfluidic pump actuated by compressive and bending force

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By

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Thin, flexible PDMS microfluidic pump actuated by compressive and bending force

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SUMMARY

Microfluidic pumps have many possible applications in the field of medine. This thesis will discuss the design and fabrication of thin PDMS microfluidic pumps that are actuated by mechanical forces resulting from compression and bending motion. The thin PDMS pumps were fabricated using replica molding and laser machining using standard PDMS mixtures to create flexible pumps. Pumps of different sizes and aspect ratios were fabricated to study the effects of geometry on the flow rate performance and the actuation force. Flow experiments were conducted to test the performance of the thin PDMS pumps. Flow rates range from 18.2 - 35.2 μL per compression and 1.8 - 11.9 μL per bend. Results indicate possible approaches to designing an efficient, simple-to-use, and electricity-free microfluidic pump.

CHAPTER 1

LITERATURE REVIEW & INTRODUCTION

Literature Review

The idea to utilize microfluidic pumps dates back to the era of development of microfluidic channels. The first ever microfluidic channel was developed in the 1970s by Stanford University. Its roots of study were based in the laboratories of Manz, Harrison, Ramsey, and Mathies.[2] However these early experiments were challenged by the nature of the material of fabrication. Being made of glass, these microfluidic channels were successful in performing DNA separation experiments but when it came to applying to protein separation, the adsorption to channel wall was problematic. New materials for fabrication emerged, including polydimethyl siloxane (PDMS) and polymethyl methacrylate (PMMA) microfluidic channels, allowing a fabrication of channels sizing from millimeters to a few microns. PDMS and PMMA were excellent substitutes for glass as they are cheaper, easier to fabricate and biocompatible.[2] PDMS and PMMA are currently popular materials used to fabricate microfluidic channels. Today numerous ways of utilizing PDMS microfluidics have emerged such as mechanics of cell separation, chemical synthesis, and microanalysis arrays.[10]

The Device

Microfluidic chips have been one of the most promising aspects in the fields of biomedical science and engineering. One of the challenges today in microfluidics is developing a better and more efficient method of delivering fluids through microfluidic channels. Currently, many electronically powered microfluidic pumps, such as Cellix Ltd and

Cole-Parmer, are in the market for the purpose of lab utility and research. These high-tech microfluidics pumps, however, have prices ranging from 1000 - 5000 US dollars and are only capable of functioning with the presence of electricity supplied by battery.[3][9] There are many applications in which battery-less pumps would be useful, for example in sub-Saharan Africa there are only 34% of hospitals that have reliable electricity access.[1][5] Our research has aimed at developing microfluidic pumps that are disposable, easy-in-use, and electricity-free. The pump consists of multiple parts: an inlet that draws in fluid, the flexible polydimethyl siloxane (PDMS) pump, and a check valve in the outlet to prohibit the reverse-flowing of the pumped liquid (shown in Fig. 1). In this thesis, we will present the first generation model of flexible, thin PDMS microfluidic pump that is facilitated by mechanical forces of compression and bending. Possible uses of our device may be to function as an electricity-free drug delivery pump for medications, medical suction to remove unwanted fluids such as pus ascites, and possibly fluid removal for soaked clothing.

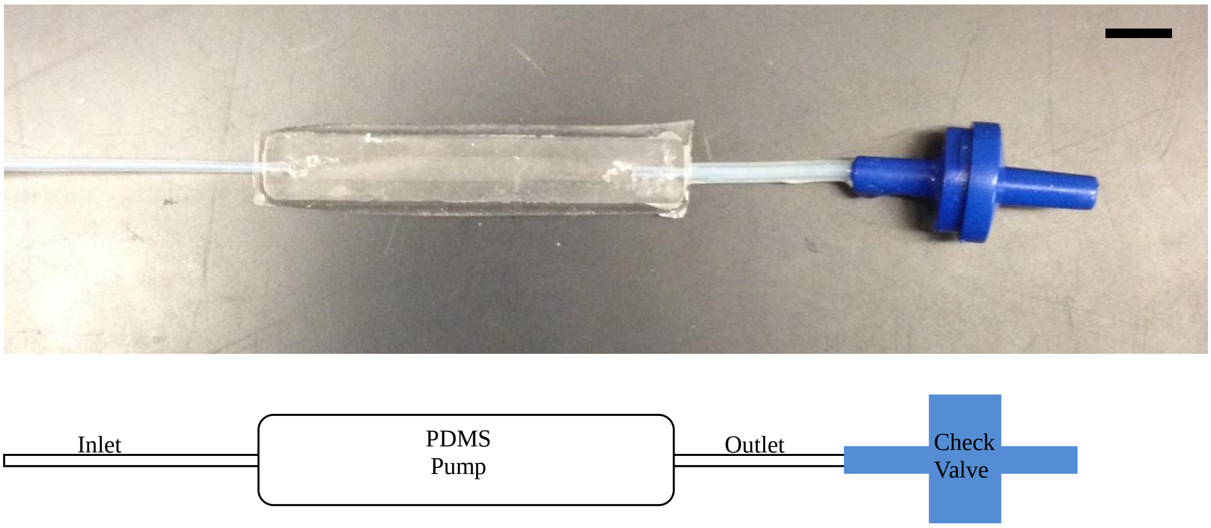


Fig. 1 Thin PDMS microfluidic pump. (Scale Bar = 1cm)

Design Origins & Research Objective

The idea of developing a thin microfluidic pump facilitated by simple mechanical forces came from the fact that the current microfluidic pumps today are inefficient and difficult to use. Many researchers have sought to design and fabricate simple microfluidic pumps that are easy to use. One of the novel microfluidic pumps developed is the "Squeeze Pump" designed by Yanyi Huang of Peking University, College of Chemistry and Molecular Engineering.[7] The device Huang *et al*. have fabricated was a simple PDMS microfluidic pump that consisted of microchannels that can be powered by a finger squeeze. The purpose of this device was to create a simple way of mixing two different chemicals within a flexible pump with few finger squeezes. However Huang's design was bulky with a size of approximately 10cm $\times$ 10cm and does not work with bending force.[7] In contrast to Huang, Valeriy Luchinov designed a self-bending microfluidic channel. His design of microfluidic channel was thinner than Huang's design and was capable of bending but only by elevating the temperature for a long period of time.[8] Our design have taken Huang's and Luchinov's fabrication and modified it to have thinner sizing, increased flexibility, and the capability of quicker pumping through bending force.

Our research objective was to investigate the functionality of thin PDMS microfluidic pumps when actuated by compression or bending force. In terms of microfluidic pump fabrication, the goal was to create a pump thin enough to be actuated by very small forces, as may exist during normal breathing. The thin pumps could be inserted within or applied on clothing or orthopedic braces, so that the device will pump in accordance with bodily movement. The microfluidic pump could be positioned near joints where most bending motions occur or across the chest where both compression and bending will occur as the person inhales. Mechanical tests were done on the device to quantify the functionality of the thin microfluidic pump.

CHAPTER 2

MATERIALS & METHODOLOGY

Microfluidic Channel Mold Preparation

A 1 inch thick extruded optically clear acrylic block was purchased from McMaster-Carr. The microfluidic channel mold was designed using Solidworks 2012 CAD software and was then converted to HAAS CNC machine language using the program HSMXpress. The converted file was transferred to a HAAS CNC machine and the mold was manufactured using a 3/64 inch flat-end mill. The molds all have a length of 2.5inches and width of 0.4inches. The channel width was designed to be 0.135 inches. Heights of the mold and channel were varied where mold height ranged from 0.2 – 0.4 inches and channel height were either 0.025 or 0.075 inches.

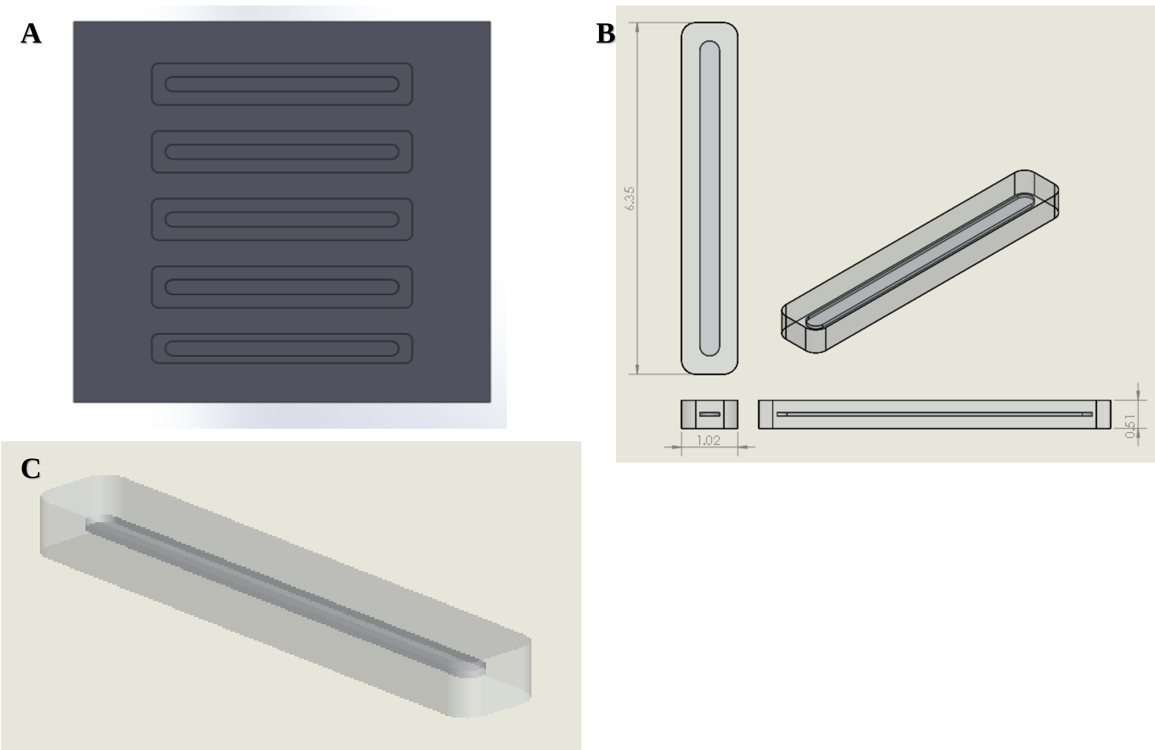


Fig. 2 CAD Models. (A) Mold design of the pumps. (B) Orthographic view of a pump. (C) Rendered transparent model of the pump.

PDMS Microfluidic Channel Preparation

Sylgard 184 silicone elastomer base and silicone elastomer curing agent were used for these experiments. The volume ratio of PDMS base to curing agent was 1:10.[4] 10mL of silicone elastomer base was extracted and well mixed with 1mL of curing agent for 10 minutes. The mixture was then poured into the mold and evacuated in vacuum chamber until all air bubbles were removed. The mixture was cured in the oven with the temperature set to 60 ℃.[4] Mixture was cured for 1 hour 30 minutes to 2 hours. After curing, finished product was left to cool in room temperature for an hour.

The channels were then glued, using the same PDMS solution, to the flat piece of cured thin PDMS. The glued pieces were cured again in the oven for 30 minutes with the temperature of 60 ℃. Once finished curing, PDMS was left to cool in room temperature.

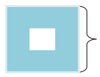
Microfluidic Pump Type	Pump Height (inch)	Channel Height (inch)	Cross Sectional Diagram
A	0.3	0.075	
B	0.4	0.075	
C	0.3	0.025	
D	0.2	0.025	

Table 1 Dimensions of microfluidic pumps. The dimensions were chosen according to the pilot study done. The best functioning pumps were chosen from the pilot study. The blue region represents the PDMS material and the white region represents the channel.

Compression Test

The fluidic device was placed flat on the lab table. Inlet tube and outlet tube were taped to prevent movement of microfluidic pump. Using both index and middle finger microfluidic pump was pressed gently on the middle section of the microfluidic pump. Microfluidic pump was compressed until 2mL of water was collected.

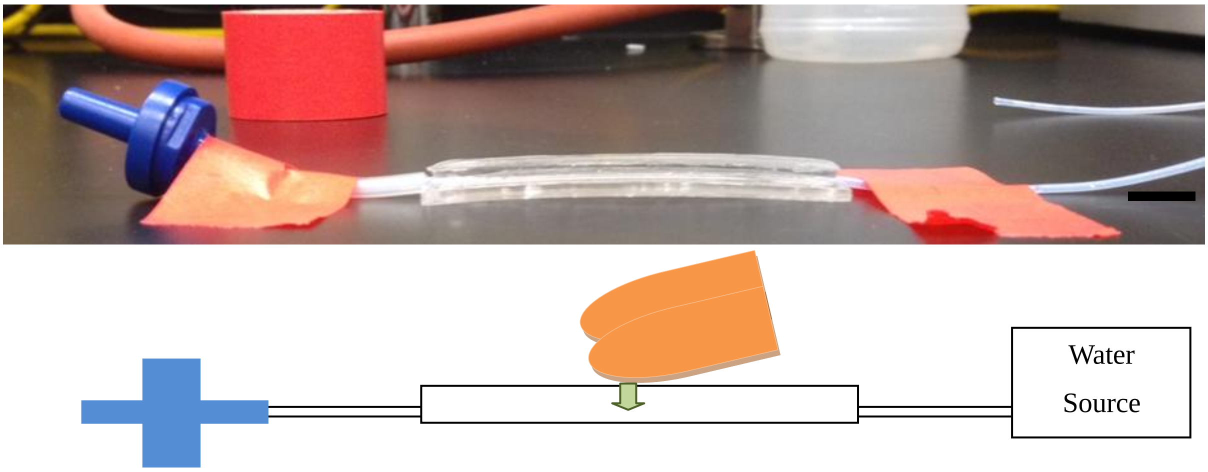


Fig. 3 Compression Testing Apparatus. (Scale Bar = 1cm)

Bending Force Test

The outlet of the microfluidic pump was taped to the lab table. A cylinderical object was placed under the mid section of the pump. Using a finger, the inlet end of the microfluidic pump is pressed downards until the end makes contact with the surface of the table. The bending is repeated until 1mL of water was collected.

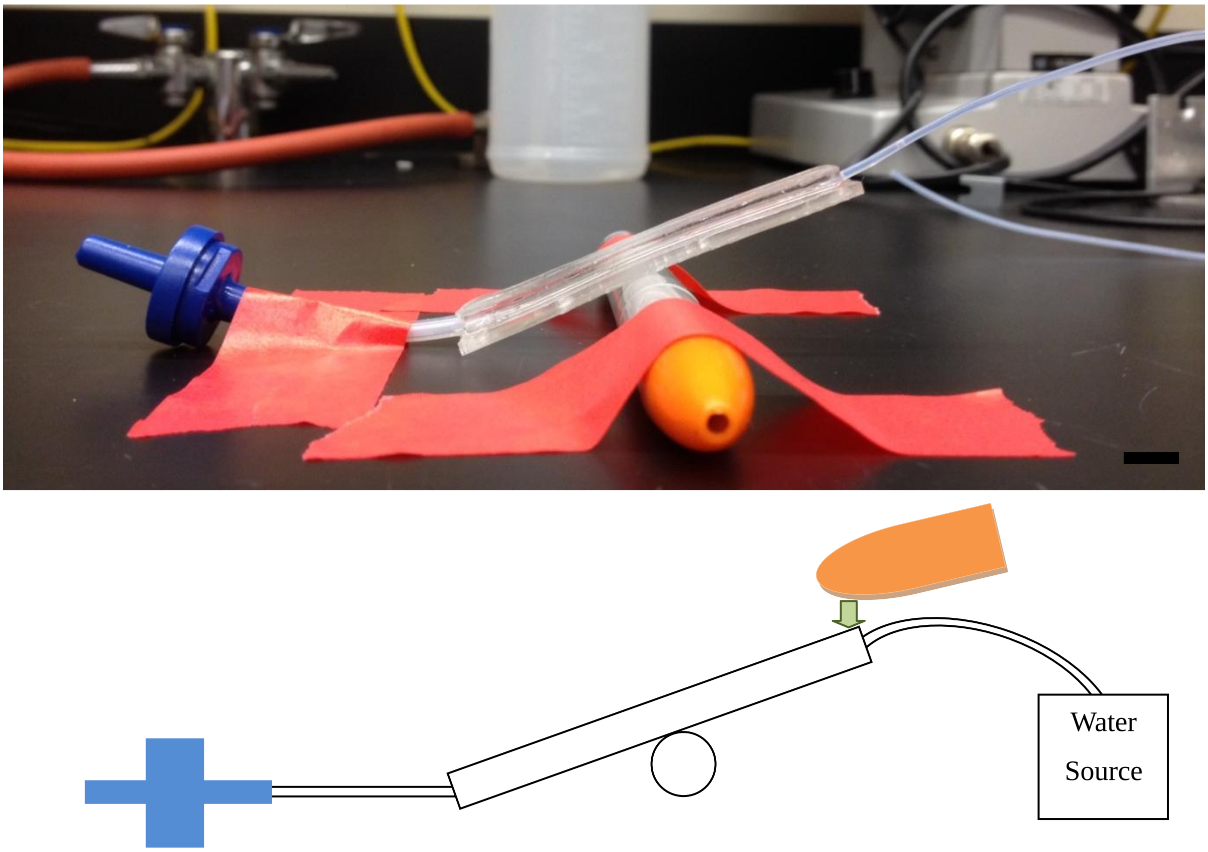


Fig. 4 Bending Force Testing Apparatus. (Scale Bar = 1cm)

CHAPTER 3

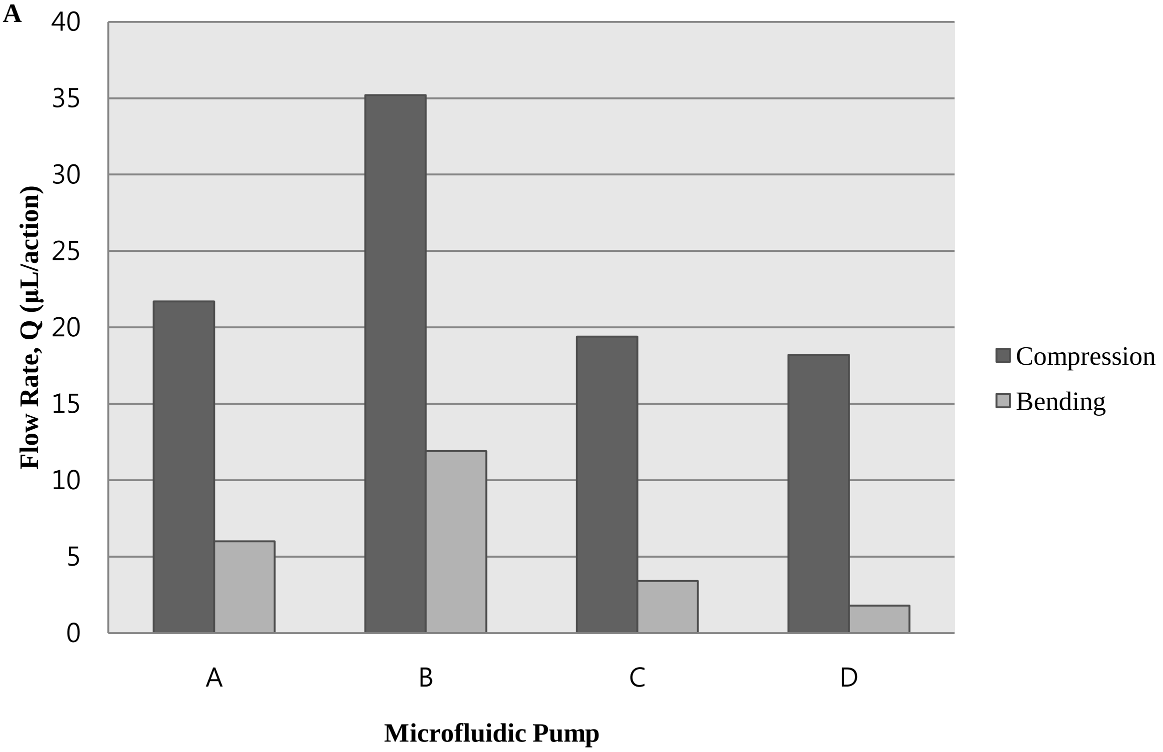
RESULTS

Multiple trials of compression and bending force testing were performed. At least 5 acceptable trials of compression and bending testing data were collected for each type of thin PDMS microfluidic pump. Any outliers that were affected by incapability of following the procedure or fatigue of fingers were neglected. The arithmetic mean was calculated for both number of compressions and number of bends (Table 2). To quantify the performance of the pump, volume of water pumped per action (compression or bending) was computed (Fig. 5).

	Type of Test	
Microfluidic Pump Type	**Compression**	**Bending**
A	92	167
B	57	84
C	103	295
D	110	560

Table 2 Mean number of compressions to pump 2mL of water and mean number of bends to pump 1 mL of water

A



B

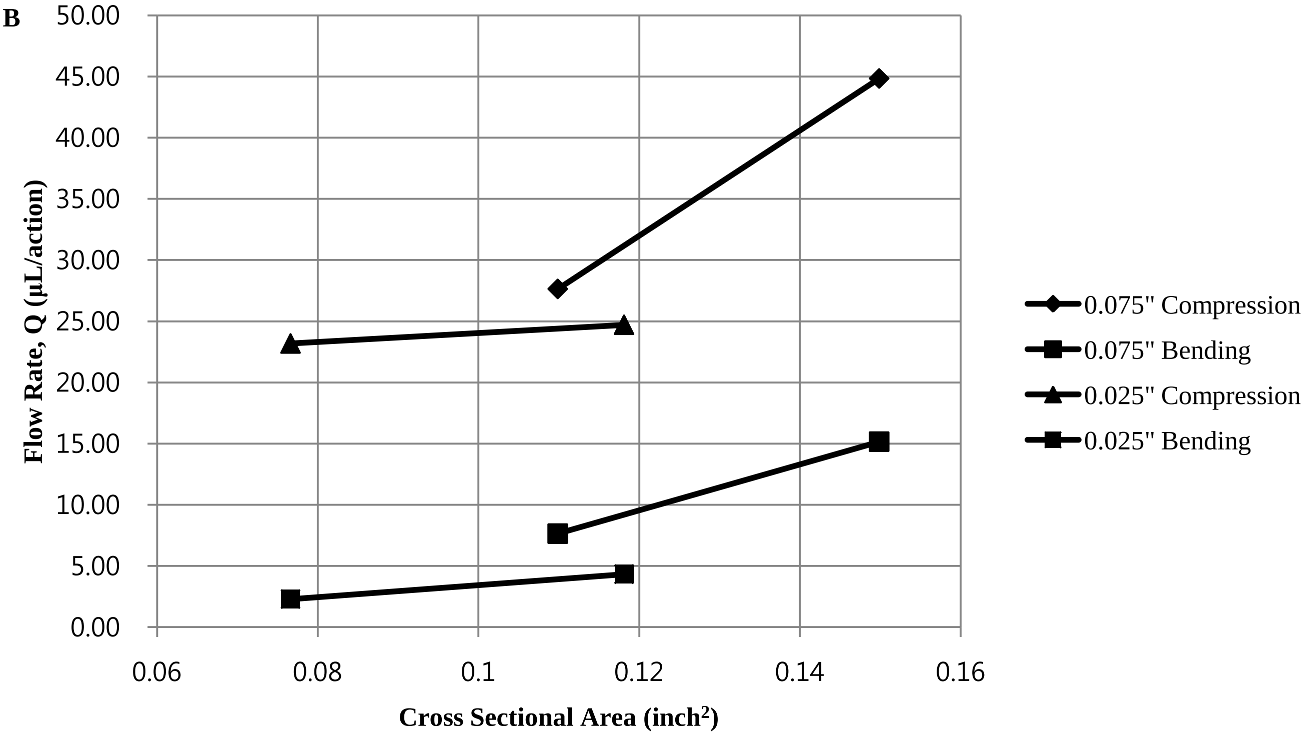


Fig. 5 Flow rate of thin PDMS microfluidic pump. (A) Compression and bending flow rate of each microfluidic pump A-D. (B) Flow rate of pumps vs. cross sectional area.

CHAPTER 4

DISCUSSION

The goal of this research was to investigate the performance of the thin PDMS microfluidic pump facilitated by the forces of compression and bending motion. In order to test its performance, flow rate was quantified for different sized thin PDMS microfluidic pumps.

Result Analysis

Looking at the flow rates of thin PDMS pump A through D, with larger pump size and thickness, flow rates are higher. The size order, in terms of thickness of the pump, is B>A>C>D. Similar trends are able to be seen for both compression and bending force test. It is clear to know that with compression, larger sized pumps have much higher flow rate than that of smaller and thinner pumps ($Q_B >> Q_A, Q_C, Q_D$). (Fig. 5B)

When Q_A, Q_C, and Q_D are compared, the difference between the flow rates are small in comparison to the difference with Q_B. Compression flow rate ratio of Q_B to Q_A is 1.6 while Q_A to Q_C is 1.11 and Q_A to Q_D is 1.2. In bending force test, flow rate ratio of Q_B to Q_A is 2.0 while Q_A to Q_C is 1.8 and Q_A to Q_D is 3.3. This indicates that the correlation between pump size and compression flow rate is lower than the correlation between pump size and bending flow rate. The reason behind this result may be due to the fact that compression correlates more to the force the pump is being compressed. On the other hand, bending is much more dependent on the deformation of the channel where higher strain of channel volume will result in higher flow.

Another relation that can be noticed is that it is more effective to increase the thickness of the pump than to increase the channel size to improve flow rate performance of the pumps. In Fig. 5A, the pump size difference between pump A and pump B is 0.1 inch while the channel sizes are the same. Flow rate differences between the A and B are 13.5 μL per compression and 5.9 μL per bend. On the other hand, there is no size difference between pump A and pump C, while the channel size of pump A is 3 times the channel size of pump C. The flow rate differences between A and C are 2.3 μL per compression and 2.6 μL per bend, which is significantly lower than the flow rate differences in the case where the pump thickness is varied. Therefore, the thickness contributes more to the performance of the pump compared to the size of the channel. However, a challenge for having thick pump is that it will take more force to deform the channel in order for the compression or bending to cause a flow. The minimal actuation force that would be produced by the body would be as low as the force caused by the expansion of chest during inhalation. The pumps can be designed to suit either of the two purposes, to have high volume flow rates or to require minimal actuation force. Increasing the flexibility of the pump material to increase the ease of deformability may reduce the minimal actuation force.

Limitations & Improvements

In compression and bending force testing, data takes into account human errors that are caused by fatiguing of fingers. Although outliers were selectively neglected from data selection, finger compressions and bends cause inconsistency in time taken for each trial of experiment. For a continuous flow of fluid through the pump, perhaps the use of solenoid actuators to compress or bend the pump could give a consistent timing of experimental trial.

Another limitation that was observed from bending force testing was the back flow of the liquid. The check valve used for the experiment is originally a product for small aquariums. It requires a certain rate of push for liquid to successfully pass through the check valves. This was noticed during bending force test when the speed of the pump bending was too slow to move the liquid forward. This indicates that not all liquid is being pumped by the bending motion is passing through the check valves. A possible improvement to this matter may be accomplished by using smaller check valves or by integrating low crackling pressure check valves into the pump itself.

CHAPTER 5
CONCLUSION

In conclusion, our device have demonstrated the thin flexible PDMS microfluidic chip to function as a pump with the facilitation of compression and bending force. The results of the research shows that increased size of pump and channel will improve performance of flow rate. The idea of thin PDMS microfluidic pump provides an alternative approach to drug delivery, medical suction and fluid transfer that is cheap, simple-in-use, disposable and electricity-free.

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